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Review Biodegradation of surgical polymers

D. F. WILLIAMS

*Department of Dental Sciences, School of Dental Surgery, University of Liverpool,
Liverpool, UK*

This work combines a review of the literature on the degradation of polymers in the physiological environment with a description of a series of experiments concerned with the role of components of that environment, such as enzymes, lipids and bacteria, on such degradation.

1. Introduction

Polymeric materials have been used in medical and surgical applications for 30 to 40 years [1-6], often in situations where there is intimate contact with living tissues. Although the specific materials requirements will differ according to the nature of the application, it is a fundamental requirement in each and every case that the polymer should display adequate biocompatibility. This implies that, for permanent implant applications, the material should not degrade within the physiological environment, nor should it have any adverse effect on the tissue, and that for short-term intentionally degradable prostheses, the rate of degradation and the release of degradation products should be physiologically tolerable.

In theory, polymeric-based materials have one significant advantage over metals, since, although the isotonic saline solution that comprises the extra-cellular fluid is extremely hostile to metals, it is not normally associated with the degradation of synthetic high molecular-weight polymers. Since the tissue response to an implanted material is often associated with the degradation of that material and the release of degradation products, this physiological inertness augurs well for good overall biocompatibility. Such a prediction is generally upheld by practical and clinical experience and it is commonly observed that the implantation of pure homo-chain high molecular-weight polymers elicits minimal response from the tissues.

In many situations this minimal response is regarded as ideal, but there are several points which deserve attention in this respect.

(a) The above generalization does not hold good for all polymers and there are some, and especially the hetero-chain polymers, which are not totally resistant to environmental degradation and have the potential for initiating a greater response.

(b) As noted above, there are some applications where it is actually desirable that the polymer degrades either in the course of its particular function (such as controlled drug release via matrix erosion) or after a specific function has been performed (such as after bone union in disposable fracture plates).

(c) Although strictly controlled, surgical polymeric materials may have to contain some additives, such as plasticizers, which themselves have a potential to irritate the tissue, especially if they are leached out from the plastic.

(d) It is not correct to assume that the physiological environment is a simple isotonic saline solution and it cannot be modelled in such a way. The extracellular fluid itself is complex, containing a variety of anions, cations and organic species, while the cells themselves may play very important parts in some reactions. It is, therefore, necessary to consider the degradation of polymers in the light of this more complex environment.

In the present work the degradation of surgical polymers is reviewed and some recent experimental work on the subject is discussed. Frequently, any degradation process that occurs in the body is referred to as biodegradation. This is not strictly correct, since biodegradation implies an active role for the biological media. Any changes produced

simply by the aqueous extracellular fluid not involving any vital function of the host, (such as simple hydrolysis) are not examples of biodegradation. Emphasis is placed in this paper on this latter aspect and a distinction is drawn between conventional degradation in a physiological environment and biodegradation.

2. Susceptibility of polymers to degradation under physiological conditions

All polymers are susceptible to degradation [7], but the conditions under which this occurs and the kinetics of the reactions are extremely variable. The degradation processes can generally be divided into two types. Firstly, there are those which involve the absorption of some kind of energy to cause disruption of primary covalent bonds to form free radicals, which then cause the propagation of molecular degradation by secondary reactions. Secondly, there are hydrolytic mechanisms where the depolymerization process can be seen as the reverse of polycondensation.

The conditions under which the first of these general processes takes place include elevated temperatures, especially in the presence of oxygen to give thermal oxidation, electromagnetic radiation (i.e., γ -rays, X-rays or ultra-violet radiation), mechanical stress at elevated temperatures giving thermo-mechanical degradation and ultrasonic vibration. Clearly, the physiological environment within the human body does not offer any of these conditions to an implanted polymer; hence, the optimistic statement that most polymers should be stable upon implantation.

Hydrolysis, on the other hand, is quite feasible in the aqueous extra-cellular fluid. A number of conditions have to be met in this respect. Firstly, the polymer has to contain hydrolytically unstable bonds. Secondly, for any significant degradation to occur, the polymer should be hydrophilic, otherwise the medium producing hydrolysis will have very limited opportunity for gaining access to the hydrolysable bonds. Thirdly, the hydrolysis has to take place at a physiological pH (around 7.4).

Thus, polymers can be placed in a ranking order of predicted susceptibility to *in vivo* degradation [8] in the sequence:

hydrophobic, no hydrolyzable bonds — most stable;

hydrophilic, no hydrolyzable bonds — may swell, but little or no degradation;

hydrophobic, hydrolyzable — surface activity only; and

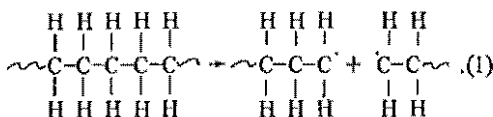
hydrophilic, hydrolyzable — bulk degradation.

The validity of these predictions is discussed below under the two headings of non-hydrolyzable and hydrolyzable polymers.

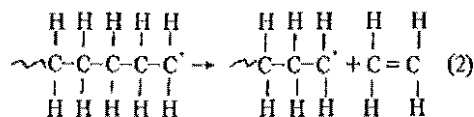
3. *In vivo* degradation of non-hydrolyzable polymers

The types of polymer used in surgery and included here are: some polyolefins (polyethylene and polypropylene), halogenated hydrocarbon polymers (most notably polytetrafluoroethylene) polyacrylic acids and their esters (such as polymethyl- and polyethyl-acrylates), some polyether urethanes and certain silicone polymers (notably polydimethyl siloxane).

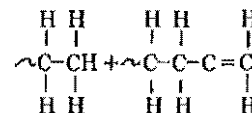
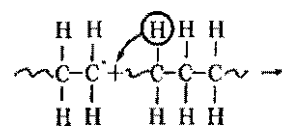
Each degradation process involves initiation, propagation and termination stages. During initiation, energy is absorbed from the external source, causing the breaking of a covalent bond (either the primary chain bond or a cross-link) and the formation of active radicals. Thus, thermal degradation occurs when the vibrational, rotational or translational energy exceeds the activation energy required to break a carbon-carbon bond on increasing the temperature. Considering the polyethylenes as examples, the thermal degradation of pure polyethylene provides one of the simplest cases, with a random chain scission initiation phase:



Propagation takes place as follows:



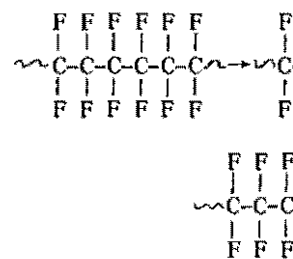
or free radical transfer may take place to terminate the process in any one molecule according to



and termination may take place by proportionation or combination

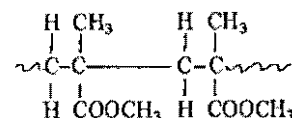
The result will be a mixture of varying sizes and some molecular transfer predominates propagation, then there will be produced. The activation energy varies with physical and chemical conditions. According to Madorsky [9], the activation energy for polyethylene of molecular weight 66,100 is 66.1 kcal mol⁻¹ and that for a polypropylene of molecular weight 11,000 is 46 kcal mol⁻¹. An ethylene will have a higher activation energy for a branched material. The activation energy for polypropylene degradation is 58 kcal mol⁻¹.

With polytetrafluoroethylene much stronger than the C-H bond, radical transfer is possible degradation. Chain scission occurs. Degradation takes place entirely by virtually 100% C₂F₄:



The activation for this reaction

The thermal degradation of polyacrylate is rather similar, involving chain scission and unzipping to yield of monomer,



hydrolyzable bonds — may swell, degradation;

hydrolyzable — surface activity

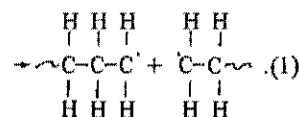
hydrolyzable — bulk degradation.

predictions is discussed below regarding the effects of non-hydrolyzable and hydrolyzable groups.

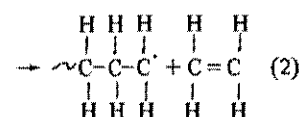
Thermal degradation of non-hydrolyzable polymers

Polymers used in surgery and included polyolefins (polyethylene and polypropylene), oxygenated hydrocarbon polymers (polytetrafluoroethylene), polyesters (such as polymethylmethacrylates), some polyether urethanes and other polymers (notably poly-

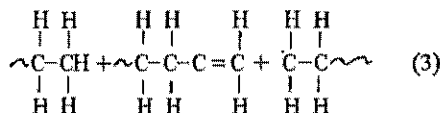
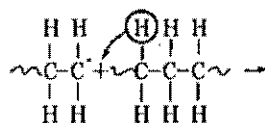
esters. The process involves initiation, propagation, and termination stages. During initiation, energy is absorbed from the external source, breaking a covalent bond (chain bond or a cross-link) to produce free radicals. Thus, thermal degradation involves the vibrational, rotational, and translational energy exceeding the activation energy to break a carbon-carbon bond on the polymer chain. Considering the principles, the thermal degradation of polymers provides one of the simplest means for chain scission initiation.



Reaction proceeds as follows:



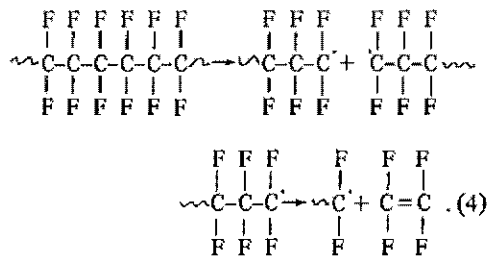
Termination may take place to terminate the reaction in one molecule according to



and termination may take place by either a disproportionation or combination process.

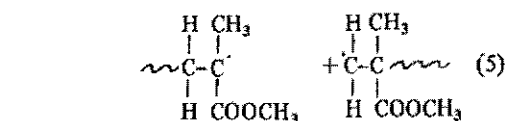
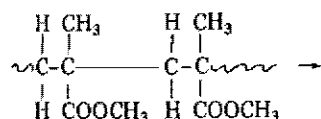
The result will be a mixture of chain fragments of varying sizes and some monomer. If the free radical transfer predominates over the unzipping propagation, then there will be little monomer produced. The activation energy for the degradation varies with physical and chemical factors. According to Madorsky [9], the activation energy for a polyethylene of molecular weight of 23 000 is 66.1 kcal mol⁻¹ and that for a molecular weight of 11 000 is 46 kcal mol⁻¹. An unbranched polyethylene will have a higher activation energy than a branched material. The activation energy for polypropylene degradation has been given as 58 kcal mol⁻¹.

With polytetrafluoroethylene the C-F bond is much stronger than the C-H bond so that no free radical transfer is possible during thermal degradation. Chain scission occurs as before, but propagation takes place entirely by unzipping, to yield virtually 100% C₂F₄:

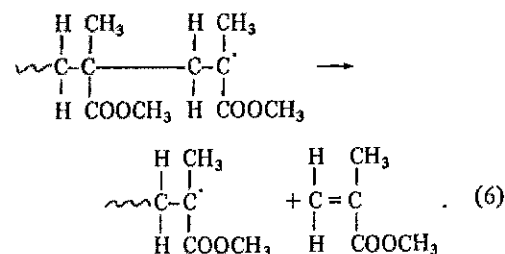


The activation for this reaction is 80.5 kcal mol⁻¹.

The thermal degradation of polymethylmethacrylate is rather similar, involving random chain scission and unzipping to yield substantial amounts of monomer,



and



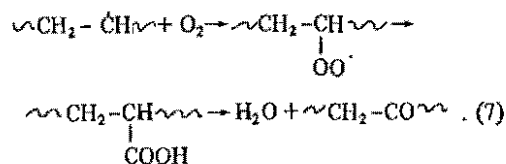
In this case the steric hindrance associated with the CH₃ and COOCH₃ groups prevents free radical transfer. The activation energy is variable, depending on the stage of the process, being 32 kcal mol⁻¹ at the beginning, rising to around 40 kcal mol⁻¹ as the reaction proceeds [10].

Radiative degradation of synthetic polymers largely involves either ultra-violet (u.v.) or high-energy radiation, the latter being defined for these purposes as radiation composed of photons or particles of higher energy than that encountered in binding electron orbitals. As discussed by Shalaby [11], the radiative degradation caused by u.v. light is called photolysis and that caused by high-energy radiation is called radiolysis. Radiolysis is far more significant than photolysis in pure synthetic polymers. The ability to degrade increases as the wavelength decreases or the energy increases. Thus, 300 to 400 nm, corresponding to an energy of 90 to 70 kcal mol⁻¹, represents the limit above which photolysis will not occur. Since radiation from sunlight involves wavelengths greater than 290 nm, and since most pure synthetic polymers will not absorb radiation of this wavelength range, u.v. radiation from sunlight is not very effective for radiative degradation. However, the presence of some impurities (such as ketones) may significantly alter the ability to absorb u.v. radiation, giving greater degradation of the polymer.

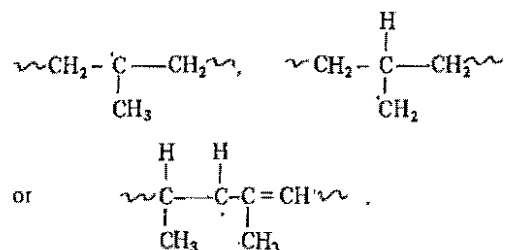
Degradation by u.v. radiation usually proceeds via the formation of free radical intermediates while degradation by high-energy radiation proceeds via the formation of both radical and ionic intermediates. The presence of oxygen in the polymer molecule has a significant influence, largely by favouring hydroperoxide formation.

Polyethylene undergoes both chain scission

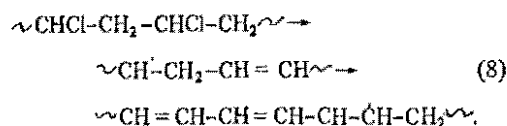
and cross-linking with the formation of a $\cdot\text{CH}$ -radical. When irradiated in air, oxidative degradation is predominant [12]:



Polypropylene may degrade via the formation of any of the free radicals



Polyvinylchloride undergoes dehydrohalogenation when either heated or irradiated:



The effects of these degradation processes will naturally vary, but generally there will be a change in average molecular weight, molecular-weight distribution, crystallinity and mechanical properties.

Chain scission will generally result in reduced strength and creep resistance, while cross-linking will be associated with an increased modulus of elasticity.

The purpose of this brief review is to show how and when degradation can occur in these polymers. Activation energies for the degradation of the high molecular-weight polymers used in surgery vary from 30 kcal mol^{-1} to 80 or 90 kcal mol^{-1} and such reactions generally require either heat, u.v. light or high energy radiation, preferably in the presence of oxygen, to proceed. It seems certain from these conditions that no such degradation should occur within the confines of the human body.

While this prediction is largely borne out in practice, there is some evidence that other factors are involved and that unexpected degradation mechanisms operate within the body. One of the

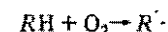
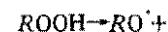
first hints of this arose from the work of Oppenheimer *et al.* [13] on the carcinogenic properties of plastics. In attempting to elucidate the mechanisms by which plastic films induced tumours after subcutaneous implantation, some radio-labelled polymers were studied in experimental animals. C^{14} -labelled polystyrene ($-\text{C}_6\text{H}_5\text{CH}-\text{C}^{14}\text{H}_2-$), polyethylene ($-\text{CH}_2-\text{C}^{14}\text{H}_2-$) and polymethylmethacrylate ($-\text{CH}_2-\text{C}(\text{CH}_3)\text{COOC}^{14}\text{H}_3-$) were used and urine, faeces and respiratory CO_2 were monitored for periods of over a year. With the polystyrene, nothing radioactive was excreted in the urine until 21 weeks, but some radioactivity was detected after this time. With polyethylene, radioactive species were excreted after 26 weeks; with polymethylmethacrylate, radioactive species were excreted after 54 weeks. Removal of the films caused the urinary radioactivity to decrease. Nothing was found in the expired air, nor in tissues surrounding the implants that were removed at sacrifice. Although the results, which showed that small amounts of degradation products were produced but that these were rapidly removed and excreted, did not help Oppenheimer *et al.* to establish mechanisms for carcinogenicity, they are very interesting from the point of view of the polymer degradation itself.

Perhaps the most significant work in this respect is that of Leibert *et al.* [14] who studied the *in vivo* degradation of polypropylene with a view to determining the degradation rate, the nature of the degradation products and the influence of antioxidant. Samples were implanted subcutaneously in hamsters. Hydroxyl concentration and carboxyl absorbance were determined as a function of implantation time. While neither the hydroxyl concentration nor the carboxyl absorbance were altered by the tissue fluids in the antioxidant-containing material, changes were seen in pure antioxidant-free polypropylene. Here the hydroxyl concentration increased linearly with time up to 100 days, after which it increased at a faster rate. Some 90 days passed before any measurable carboxyl was formed. Gas-phase chromatography (GPC) analysis showed a slight shift in molecular-weight distribution, with a decrease in the proportion of very large molecules and an increase in the number of medium-sized molecules. Dynamic mechanical testing showed a slight decrease in $\tan \delta$ during the first 40 days for the pure specimens, where δ is the phase angle between the applied strain and the resultant stress.

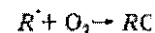
The authors attempted to date the degradation with oxidative degradation. It has been suggested that the energy for the initiation step is 31 kcal mol^{-1} and there is a function of temperature and concentration. Hydroperoxide



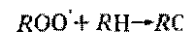
and when these decompose, causing carboxyl groups to form



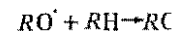
and



Rapid oxidation then follows in carboxyl content and loss. During propagation, both hydroperoxide and carboxyl content increase:



and



At 75°C and in 100% oxygen, polypropylene degrades by the above mechanism. Extrapolation of data to 3 logical oxygen partial pressures predicts an hydroxyl production of $10^{-4} \text{ (mg OH) (g polymer)}^{-1}$ per year. Leibert *et al.*, degradation occurred at a much faster rate, with a period of about 100 days. trace amounts of metallic species could be responsible for the oxidation rate.

The suggestion is made, that it may be influential in degradation, in fact, have been made, but always without proof. Leibert speculated that degradation caused by enzymatic oxidation. He quoted some unpublished data in polyethylene film were degraded over weeks by exposing them to enzymes which produce relatively large amounts of carboxyl groups.

There is a certain attraction for the suggestion of enzyme-accelerated degradation, since enzymes have been found to accelerate the degradation of polymers.

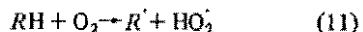
ose from the work of Oppenheimer on the carcinogenic properties of polypropylene implants to elucidate the mechanical films induced tumours after implantation, some radioisotopes were studied in experimental animals. Polystyrene ($-C_6H_5CH-C^{14}H_2-$), polyethylene ($-C^{14}H_2-$) and polymethylmethacrylate ($-C(CH_3)COOC^{14}H_3-$) were used and respiratory CO_2 was analysed for over a year. With the polypropylene, no radioactivity was excreted in the first 26 weeks, but some radioactivity was excreted after 26 weeks; with polyethylene, no radioactivity was excreted after 26 weeks; with polymethylmethacrylate, radioactive species were excreted after 54 weeks. Removal of the polypropylene radioactivity to decrease in the expired air, nor in the implants that were removed. The results, which showed that the degradation products were removed and help Oppenheimer *et al.* to conclude for carcinogenicity, they concluded from the point of view of the polypropylene itself.

The most significant work in this field is by Liebert *et al.* [14] who studied the degradation of polypropylene with a view to the degradation rate, the degradation products and the influence of the implants. Samples were implanted in rats. Hydroxyl concentration and absorbance were determined after implantation time. While neither the concentration nor the carboxyl concentration increased linearly with time, after which it increased at a rate of 1.90 days passed before any carboxyl was formed. Gas-phase mass spectrometry analysis showed a slight shift in weight distribution, with a portion of very large molecules. The number of medium-sized molecules was determined by mechanical testing showed a change in δ during the first 40 days of implantation, where δ is the phase angle of strain and the resultant stress.

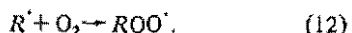
The authors attempted to compare this degradation with oxidative degradation at high temperatures. It has been suggested that the activation energy for the initiation step in this oxidation is 31 kcal mol⁻¹ and there is an induction time which is a function of temperature and oxygen concentration. Hydroperoxides are formed:



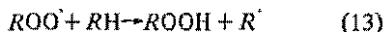
and when these decompose, chain scission occurs, causing carboxyl groups to form:



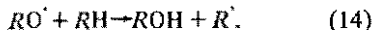
and



Rapid oxidation then follows, causing an increase in carboxyl content and loss of tensile properties. During propagation, both hydroxyl and hydroperoxide content increase:



and



At 75°C and in 100% oxygen, polypropylene degrades by the above mechanism [15] and extrapolation of data to 37°C under a physiological oxygen partial pressure, pO_2 , of 15 mm Hg predicts an hydroxyl production rate of 2.9×10^{-4} (mg OH) (g polymer)⁻¹ day⁻¹ and an induction period of 20 years. Clearly, in the work of Liebert *et al.*, degradation under these conditions occurred at a much faster rate, with an induction period of about 100 days. They speculated that trace amounts of metallic ions, enzymes or other species could be responsible for this increased oxidation rate.

The suggestion is made, therefore, that enzymes may be influential in degrading polymers. Several authors, in fact, have made this suggestion, but always without proof. Leininger [16], for example, speculated that degradation of polyethylene was caused by enzymatic oxidative chain scission and quoted some unpublished work where changes in polyethylene film were found in a matter of weeks by exposing them to bacterial cultures which produce relatively large amounts of oxidative enzymes.

There is a certain attractiveness in the hypothesis of enzyme-accelerated *in vivo* polymer degradation, since enzymes have the characteristic

ability to catalyse certain chemical reactions; it would certainly be very useful to invoke this hypothesis in the case, above, of polypropylene to explain why the degradation, which involves a high activation energy, occurs at room temperature. There are some difficulties involved in confirming the hypothesis, however, since enzymes are normally so substrate specific and one does not normally associate their catalytic effect with synthetic high molecular-weight polymers. Nevertheless, it remains an attractive possibility, worthy of further investigation.

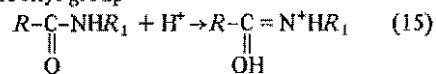
The possible role of enzymes is discussed at greater length in the following section on hydrolyzable polymers.

3. Hydrolyzable polymers

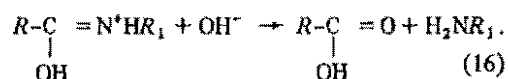
Hetero-chain polymers, particularly those containing oxygen and/or nitrogen atoms in the main chain, are generally susceptible to hydrolysis. Depending on the structure, this hydrolysis may be favoured by either acid or alkaline conditions and naturally is much faster at elevated temperatures. Although hydrolysis of many of these polymers will therefore be most marked under, for example, hot alkaline conditions, the aqueous environment of the body at 37°C is sufficiently hostile to degrade a number of polymers by hydrolysis. It is in this context that enzymes, and especially hydrolytic enzymes, are most likely to have an effect. Amongst the polymers which have been shown to degrade by hydrolysis *in vivo* are certain polyamides, including nylons and polyamino acids, some polyurethanes, cyanoacrylates and some polyesters, both aromatic and aliphatic.

3.1. Nylon

Some 25 years ago, Harrison [17, 18] showed that nylon fabrics lost about 80% of their tensile strength during a three-year implantation. Leininger *et al.* also found considerable degradation of nylon [19]. Some nylons are hydrophilic and hydrolyzable, although the extent of water absorption is variable. Nylon 6, for example, has a water content at saturation of 11%, while nylon 11 has a water content of only 1.5%. The hydrolysis mechanism is quite simple [7], the primary attack being that of the hydrogen ion on the oxygen atom of the carboxyl group



and



Thus, acid and amine end groups are formed. Very few studies have been performed on the actual *in vivo* degradation of nylons, although Williams has found that nylon 6-6 degrades faster in the tissues of an acute inflammatory response than in the more quiescent chronic phase of a tissue response [20].

Several reports have also recently been published concerning the unexpected and undesirable degradation of nylon sutures used to secure ophthalmological prostheses in place [21, 22].

3.2. Polyamino acids

Synthetic polyamino acids and polypeptides of more complex main-chain and pendant group structure are often susceptible to hydrolysis. It is now well-known that these materials may be degraded by enzymes *in vitro*, where the enzyme substrate specificity in relation to the bonds broken is unchanged. Much data is available showing the solubility and digestibility of polyamino acids by proteolytic enzymes [23]. For example, synthetic polylysine has been studied on numerous occasions. Trypsin catalyses the degradation of this polymer [24], the chief products being dilysine and trylisine. Interestingly, no lysine is produced, indicating that the terminal peptide bonds are not attacked. This is in contrast to the acid hydrolysis of polylysine where random scission degradation occurs, giving lysine amongst the products.

Further evidence of the role of enzymes on the degradation of polyamino acids *in vivo* and *in vitro* has recently been supplied by Dickinson and co-workers [25, 26]. Several proteolytic enzymes were employed *in vitro* to assess the degradability of cross-linked poly(2-hydroxyethyl-L-glutamine). Trypsin and collagenase had no effect but pronase and papain dissolved the hydrogel. Analysis of the papain digestion products showed that mainly oligomers of degree of polymerization 4 to 9 were produced. It was suggested that trypsin and collagenase were too substrate specific. Degradation of this material was also observed *in vivo*, although this was confined to the first two weeks, as judged by changes in the swelling characteristics. It was suggested that, since this two weeks corresponded with the maximum cellular

activity, it was enzymes released from the cells of the wound-healing response that were responsible. The cellular layer at the tissue-implant interface in the chronic response was associated with little or no degradation activity. These conclusions are very interesting in the light of the work of the author discussed later in this paper.

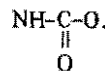
An interesting and complicating factor here is the inhibition of enzyme activity when low molecular-weight peptides are released into enzyme solutions at low concentrations. Thus, while degradation occurs with polyamino acids of molecular weight greater than 40 000, with low molecular weights in the region of 2 500, the enzyme and polymer may form an insoluble complex. This is an important point in the interpretation of *in vitro* work on polyamino acid degradation.

Subtilisin, α -chymotrypsin, ficin, papain and elastase also degrade polylysine, in that order of reactivity [27]. Other polymers of α -amino acids that are degraded by hydrolytic enzymes include poly-DL-alanine, poly-L-aspartic acid and poly-L-proline.

The hydrolytic instability of the amide bond in synthetic amino acid polymers has been used in the formation of intentionally biodegradable polymers. As reviewed by Kopecek [28], these polymers may be designed to give controlled biodegradation via the introduction of segments susceptible to effects by specific enzymes. For example, polymers of *N*-(2-hydroxypropyl) methacrylamide and *p*-nitrophenyl esters of *N*-methacryloylated amino acids are reacted with compounds containing an aliphatic amino group, with formation of the amide bond. If this bond originates in an amino acid specific for a certain enzyme, an enzymatically cleavable bond is formed. Specific acids for chymotrypsin, for example, and L-phenylalanine, L-tyrosine and L-leucine. A wide variety of degradable polymers has been prepared in this way.

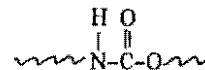
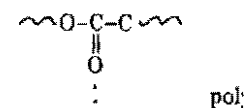
3.3. Polyurethanes

Polyurethane is the name given to a group of polymers containing the urethane group,

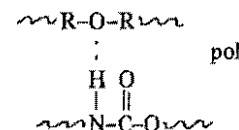


A wide variety of polyurethanes exist in which different groups are present in adjacent molecular chains, most commonly involving urethane, urea,

ester or ether groupings, with through hydrogen bonding. They are classified according to whether polyester or polyether in nature

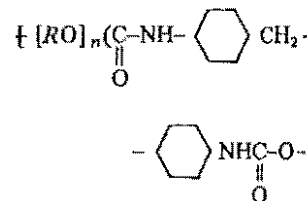


or

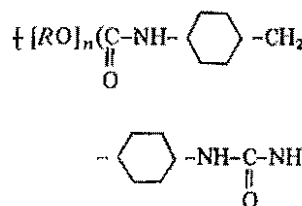


The polyester urethanes tend to be unstable due to their highly flexible configurations. Unfortunately, uses of polyurethanes involving *in vivo* degradation are common [29-31]. Mirkovitch polyester urethane used as an artificial blood vessel was found to be considerably degraded during a six-month period.

On the other hand, the polyether urethanes are far more stable. Of particular interest is segmented polyether urethane



and the segmented polyether urethane

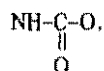


which are flexible stable materials that withstand long implantation without noticeable sign of degradation

zymes released from the cells
ig response that were respon-
layer at the tissue-implant
ronic response was associated
degradation activity. These
interesting in the light of the
discussed later in this paper.
nd complicating factor here is
enzyme activity when low
ptides are released into enzyme
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in the region of 2500, the
ner may form an insoluble
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ro work on polyamino acid

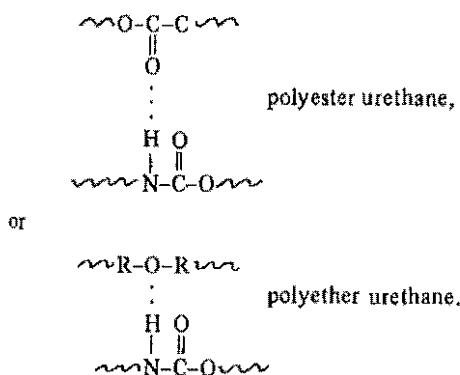
motrypsin, ficin, papain and
de polylysine, in that order
Other polymers of α -amino
raded by hydrolytic enzymes
alanine, poly-L-aspartic acid
instability of the amide bond
acid polymers has been used
of intentionally biodegradable
wed by Kopecek [28], these
designed to give controlled
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s of *N*-(2-hydroxypropyl)
nd *p*-nitrophenyl esters of
1 amino acids are reacted
containing an aliphatic amino
ion of the amide bond. If this
an amino acid specific for a
enzymatically cleavable bond
acids for chymotrypsin, for
phenylalanine, L-tyrosine and
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es
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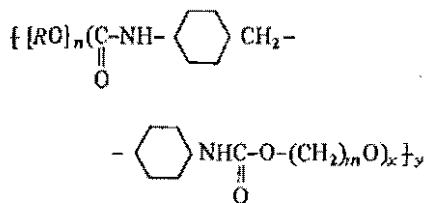
polyurethanes exist in which
e present in adjacent molecular
only involving urethane, urea,

ester or ether groupings, which are combined
through hydrogen bonding. They are generally
classified according to whether they are primarily
polyester or polyether in nature:

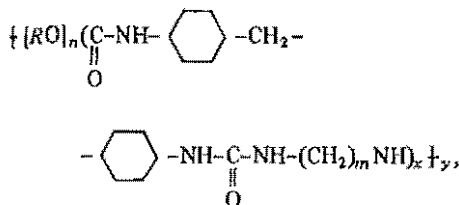


The polyester urethanes tend to be hydrolytically
unstable due to their highly strained molecular
configurations. Unfortunately, the early surgical
uses of polyurethanes involved these varieties
and *in vivo* degradation and disintegration was
common [29-31]. Mirkovitch *et al.* [32] studied
polyester urethane used as an aortic graft in dogs
and found that the molecular structure changed
considerably during a six-month implantation
period.

On the other hand, the polyether urethanes are
far more stable. Of particular interest here are the
segmented polyether urethanes,



and the segmented polyether urethane ureas,



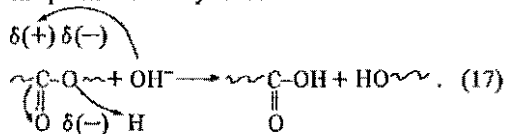
which are flexible stable materials that appear to
withstand long implantation times without any
noticeable sign of degradation.

3.4. Cyanoacrylates

The degradation of these surgical adhesives and
hemostasis agents was thoroughly investigated
by Leonard [33,34]. Using radioactive poly-
methyl-2-cyanoacrylate, contained in a poly-
vinyl alcohol sponge, they showed that, after
seven days, 0.2% of the radioactivity had been
excreted in the faeces and 8.5% had been excreted
in the urine, with 85.4% remaining in the sponge
and 5.9% unaccounted for. After 159 days only
6.6% remained in the sponge, 45.1% having been
removed in the urine and 4.3% in the faeces, the
remaining 44% presumably being expired as CO₂.
In vitro, the degradation appeared to be initiated
by OH⁻ ions and produced formaldehyde and an
alkylcyanoacetate after a random hydrolytic
chain scission mechanism of propagation. Vezin
and Florence [35] have recently studied further
the *in vitro* heterogeneous degradation of
poly(*n*-alkyl α -cyanoacrylates) and found that
the degradation mechanisms proposed by Leonard
et al. [33] may not always apply but may, depen-
ding on molecular weight and molecular-weight
distribution, involve initiation at chain ends.
Enzyme activity does not appear to have been
invoked with these polymers.

3.5. Polyesters

The ester bond is readily hydrolyzed, resulting
from the primary attack of the hydroxyl ion on
the positive carbonyl C-atom:



The susceptibility of individual polyesters
depends on specific molecular structures and
hydrophilicities. In general, the aromatic poly-
esters are less sensitive to moisture than the
aliphatic polyesters because of the greater hydro-
phobicity of the aromatic parts.

3.6. Aromatic polyesters

Polyethylene terephthalate (PET) is the most
relevant polyester to mention here, being
widely used for prosthetic devices. Although
Harrison and co-workers showed some degradation
of a woven PET (Dacron) after a three-year implan-
tation [17,18] it is generally considered suf-
ficiently hydrophobic to render bulk degradation
unlikely. Surface degradation may occur slowly,

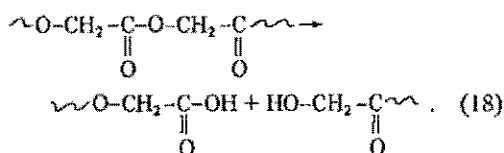
but usually there is little change in mechanical properties over a long period of time [36].

The kinetics of PET degradation have been recently discussed by Rudakova *et al.* [37]. They implanted samples subcutaneously in rabbits and dogs and found the time for complete degradation was 30 ± 7 years. Fifty per cent of the initial strength of PET filaments was lost in 10 ± 2 years.

3.7. Aliphatic polyesters

Aliphatic polyesters, typified by polyglycolic acid and polylactic acid, have been studied extensively, since they provide model systems for investigating biocompatibility and also provide useful materials for applications requiring controlled degradation, such as synthetic absorbable sutures.

Polyglycolic acid is the most hydrophilic of all polyesters [38], hydrolysis readily occurring via the ester bond, yielding alcohol and carboxylic acid groups:



Reed and Gilding [39] have shown that the loss of tensile strength of polyglycolic acid sutures (Dexon®) is quite rapid when tested *in vitro* at pH 7 and 37°C, with total loss of strength occurring after 4 to 6 weeks. This has been confirmed by Chu [40] and Williams [41]. Reed and Gilding [39] also found that the *in vivo* degradation was similar to *in vitro* degradation, and could, in fact, be modelled *in vitro* by using a 0.2M phosphate buffer at pH 7 and 37°C. Although the tensile strength is reduced to zero after 32 days, the mass-loss does not begin until 21 days, the oligomeric fraction formed by degradation becoming the main component. The conclusion that the *in vivo* degradation of polyglycolic acid is a result of simple hydrolysis is not totally substantiated by the work of the author, as discussed below. Gilding [38] has stated that the hydrolysis takes place preferentially in the amorphous phase of this semi-crystalline material, and this is supported by the work of Chu [40] who has argued that water is able to penetrate amorphous areas more rapidly than crystalline areas and tie-chain segments in these

regions begin to degrade. When the amorphous regions have been removed by hydrolysis, the second stage of degradation starts, thus involving the crystalline areas. The degree of crystallization was observed to increase during the first stage of degradation and then to decrease during the second state.

Other aliphatic polyesters of the same series, and their co-polymers, have been shown to become degraded by hydrolysis, although at substantially slower rates [39]. Further comments on the role of enzymes and bacteria in the degradation of these polymers are given below.

4. Experimental observations on the role of the physiological environment in polymer degradation

In view of the conjecture and discussion concerning the precise mechanisms by which polymers degrade *in vivo*, a series of experiments have been carried out in which certain aspects of the role of the physiological environment in polymer degradation have been studied.

4.1. The effect of enzymes on polyglycolic acid [41, 42]

It was considered important to identify whether enzymes, under any conditions, were able to degrade synthetic high molecular-weight polymers, this hitherto only having been demonstrated in the case of some polyamino acids. Polyglycolic acid was chosen for the initial studies, in which Dexon sutures were incubated in various solutions, since this polymer is hydrolyzed in aqueous media, so that the rates of hydrolysis could be compared in the presence or absence of enzymes. The enzymes studied were acid phosphatase, bromelain, carboxypeptidase-A, chymotrypsin, clostridiopeptidase-A, esterase, ficin, leucine aminopeptidase, papain, peptidase, pepsin, pronase, proteinase-K and trypsin. The sutures were immersed in solutions of these enzymes in appropriate buffers for varying periods of time at 37°C and the degradation, monitored mechanically, was compared to that produced in buffers alone.

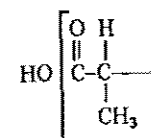
Acid phosphatase, papain, pepsin, peptidase, pronase, proteinase-K and trypsin had no apparent effect on the polymer. Ficin, carboxypeptidase-A, chymotrypsin and clostridiopeptidase-A all produced significantly greater amounts of degradation, often increasing the rate of hydrolysis by a factor of two. Bromelain, esterase and leucine amino-

peptidase all had very significant effects. It was difficult to take into account the amount of degradation due to sulphate, present in the solution, present in the solution.

While the varying activities and the different amounts of activity during the experiment, a comparative and quantitative effect of these enzymes, it was under some conditions, are polymer degradation. It is into the enzymes that did influence mainly (although not exclusively) might be expected to attack a on the basis of its molecular structure.

4.2. The effect of enzyme acid [43]

Similar studies have been carried out on lactic acid, of structure



although in this case powder instead of multi-filament sutures was monitored analytically. Of the enzymes, dehydrogenase gave entire. No lactic acid was detected. Exposure of polylactic acid was a slight weight-loss and found that there was a slight attack without yielding any low molecular-weight degradation effect, judged by a small weight of lactic acid was noted with.

On the other hand, when employed, pronase, proteinase all had a significant effect. In three cases, the polymer weight with the detection solution, by both thin-layer and a qualitative test with pH also a concomitant reduction action may not have been the cases, however, since both caused a physical break-down.

grade. When the amorphous removed by hydrolysis, the radiation starts, thus involving. The degree of crystallization increase during the first stage of then to decrease during the

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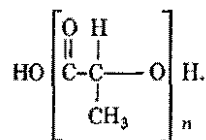
ase, papain, pepsin, peptidase, e-K and trypsin had no apparent mer, ficin, carboxypeptidase-A, l clostridiopeptidase-A all pro- greater amounts of degradation, ie rate of hydrolysis by a factor n, esterase and leucine amino-

peptidase all had very significant effects, although it was difficult to take into account quantitatively the amount of degradation due to the ammonium sulphate, present in the solution to stabilize the enzymes.

While the varying activities of these enzymes and the different amounts by which they lose activity during the experimental period preclude a comparative and quantitative analysis of the effect of these enzymes, it was clear that enzymes, under some conditions, are able to influence polymer degradation. It is interesting to note that the enzymes that did influence the hydrolysis were mainly (although not exclusively) of the type that might be expected to attack an aliphatic polyester on the basis of its molecular structure, i.e., esterases.

4.2. The effect of enzymes on polylactic acid [43]

Similar studies have been carried out on poly-L-lactic acid, of structure



although in this case powdered polymer was used instead of multi-filament sutures and the degradation was monitored analytically, rather than mechanically. Of the enzymes studied, lactate dehydrogenase gave entirely negative results. No lactic acid was detected in solution after exposure of polylactic acid to esterase, but there was a slight weight-loss and fall in pH; it is possible that there was a slight attack on the polymer without yielding any low molecular-weight fractions. Ficin also gave equivocal results with the suggestion of a slow action yielding relatively high molecular-weight degradation products. A slight effect, judged by a small weight-loss and traces of lactic acid was noted with trypsin.

On the other hand, under the conditions employed, pronase, proteinase-K and bromelain all had a significant effect on this polymer. In all three cases, the polymer showed a reduction in weight with the detection of lactic acid in the solution, by both thin-layer chromatography and a qualitative test with p-hydroxydiphenyl, and also a concomitant reduction in pH. The mode of action may not have been the same in these three cases, however, since both pronase and bromelain caused a physical break-down in the polymer,

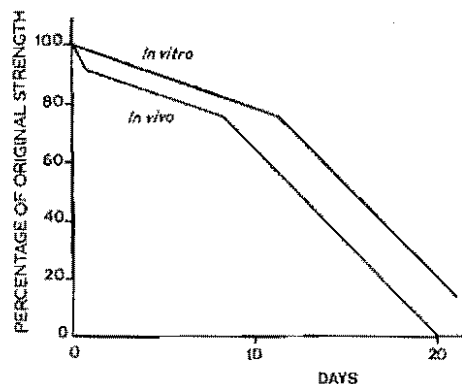


Figure 1 *In vivo* and *in vitro* profiles for the degradation of polyglycolic acid sutures [41].

giving a much finer dispersion, while proteinase-K did not substantially alter the physical form of the powder, although there was a similar weight-loss and production of lactic acid. It is conceivable that these enzymes are displaying a different activity in these cases: perhaps the pronase and bromelain exhibiting exokinase behaviour and the proteinase-K exhibiting endokinase behaviour.

4.3. *In vivo* degradation of polyglycolic acid [41]

The degradation of the polyglycolic acid sutures (Dexon[®]) was studied *in vivo* by the subcutaneous implantation of specimens in rats, monitoring the degradation mechanically. *In vitro* tests were performed in parallel with the *in vivo* tests using Tris buffer at pH 7.4. The degradation profiles are given in Fig. 1, from which it is clear that the material does not display similar behaviour *in vivo* and *in vitro*, as claimed by Reed and Gilding [39]. A considerably greater number of time intervals were chosen for study in the present case, giving an insight into the difference between the two conditions. *In vitro* it is apparent that the rate of loss of strength is linear for about 12 days, by which time a 25% loss has been recorded. At this point the slope changes, the rate of loss becoming more rapid, although still linear, until disintegration between 25 and 30 days. *In vivo* at every time interval studied the strength was less than that recorded *in vitro*. However, this was due to a rapid initial loss of strength, occurring within the first two days. This observation was checked very carefully by subsidiary experiments in which specimens were implanted for short

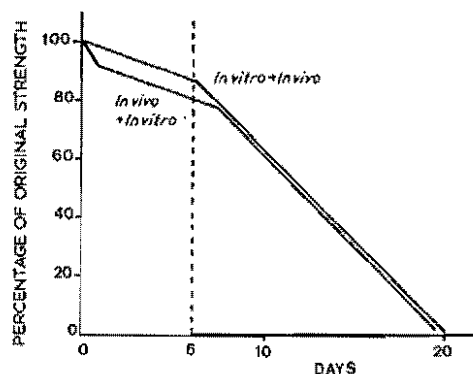


Figure 2 Degradation profiles for polyglycolic acid sutures, with transfer between *in vivo* and *in vitro* conditions at 6 days [41].

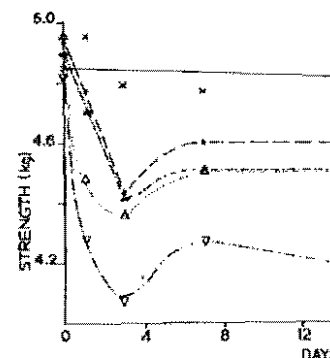
period of time, the rapid drop being confirmed. After two days, the suture loses strength at the same rate as those in the first stage of *in vitro* degradation, until the same 25% loss of strength has been reached, in this case, at about 8 days. At this point the rate increased, again to an identical value as the *in vitro* specimens.

These observations would suggest that either the initiation of degradation occurs far more readily in the *in vivo* environment than in a buffer solution but that, once initiated, the degradation proceeds by the same hydrolytic mechanism whatever the environment, or that there is something very specific about the first few days *in vivo* that accelerates the degradation process. This was checked by experiments in which some specimens were kept in buffer for 6 days and then implanted, while other specimens were implanted for 6 days and then removed and kept in buffer. Those which were transferred from *in vivo* to *in vitro* conditions followed the same degradation profile throughout as the previous *in vivo* specimens (see Fig. 2), but those which were transferred from *in vitro* to *in vivo* conditions followed the *in vitro* profile at the beginning, but then suffered a rapid loss of strength on implantation, thereafter following the *in vivo* profile. This would indicate that there is something specific about the immediate environment on implantation that influences the hydrolysis mechanism, at any stage in the degradation process. The fact that the profiles are so similar, apart from the period around the implantation time, however, would suggest that at other times *in vivo*, the degradation process is similar under the two con-

ditions, and the results are quite in keeping with the hypothesis of Chu, discussed earlier [40]. These results are also consistent with those of Dickinson and co-workers [25, 26] discussed earlier, where the *in vivo* degradation of polyamino acids was largely confined to the acute phase of the tissue response.

What it is in the immediate post-implantation physiological environment that has this effect is a matter of speculation. It is tempting to assume that it is enzymes released from cells in response to the trauma of implantation that are responsible, but it is very difficult to prove or deny this. An attempt has been made to investigate this by using a novel technique in which specimens are maintained in varying types of inflammatory tissue. This is based on the hypothesis that, if enzymes are able to influence polymer degradation, then, since different enzymes are being synthesized and released at different rates during the various phases of the tissue response to an implant, the polymer will degrade non-linearly as a function of time. It may be, for example, that the enzymes released by cells, such as polymorpho-nuclear leucocytes, that dominate in the first few days, are the most active in this respect, in which case there will be a rapid initial degradation, followed by a slower rate (ignoring for the moment any variation in degradation rates associated with structural factors). On the other hand, should the polymer be more susceptible to the enzymes released by the macrophages of the chronic response, then the rate would increase with time.

In this particular technique, specimens were implanted in rats, the samples being divided into two groups. Those of the first group were left for the duration of the experiment and checked for degradation on removal. Those of the second group were removed from the animals after a short period of time and re-implanted into fresh animals. This process was repeated at appropriate times during the experiment. In the first group, the specimens experienced an acute response followed by a chronic response, while those in the second group experienced a series of acute responses but were never subjected to the chronic phase. Although, as discussed below, this technique has given very interesting results with other polymers, in the case of polyglycolic acid, in which the second group were re-implanted four times in an eleven-day period, very little change was noted.



The re-implanted specimens had more degradation (55.8% strength compared to 59.9%) which was significant at $p < 0.005$, but not a conclusive difference. This was due to the fact that the time period is too short to be a significant change in inflammatory tissue and tend to observe that the propagating hydrolysis is largely independent of the environment. The attention has now been turned to other more prolonged degradation profiles which will give better results.

Enzymes, of course, are one of the species of the physiological environment that could affect polymers, and it is considered the alternatives. On the other hand, the alternatives are the lipids present in fluids. It is well-known that lipids influence the structural integrity of materials, most notably in heart valves in which of this material have malfunctioned. Structural changes in the rubber-silicone rubber joint prostheses are apparently because of environmental interactions. Although the extent of degradation are not entirely certain, it is certain that lipids and other high molecular weight substances are involved.

Some *in vitro* experiments have been performed in which polyglycolic acid was exposed to varying concentrations of lipids: butyric acid, caproic acid, and stearic acid [48]. Figure 3 shows a significant loss of strength of samples in a few days. Interestingly, samples that were re-implanted suffered little change in strength and suffered little

ults are quite in keeping with Chu, discussed earlier [40], also consistent with those of workers [25, 26] discussed *in vivo* degradation of polyamino confined to the acute phase of

the immediate post-implantation environment that has this effect. It is tempting to assume that released from cells in response to implantation that are responsible, but to prove or deny this. An experiment was made to investigate this by a technique in which specimens are of varying types of inflammatory response based on the hypothesis that, if different enzymes are being released at different rates during the tissue response to an implant, they will degrade non-linearly in time. It may be, for example, that released by cells, such as polymorphonuclear leukocytes, that dominate in the early phase of the response, are the most active in this respect, and will be a rapid initial degradation followed by a slower rate (ignoring for variation in degradation rates structural factors). On the other hand, if the polymer be more susceptible to degradation by the macrophages of the late phase, then the rate would increase

over a period of time in these particular solutions (in a phosphate buffer at pH 7.0). This subsequent rise in strength may be associated with an absorption of lipids into the polymer or associated swelling. Fig. 4 shows the relationship between the time taken for a suture to break at a nominal load of 2.55 kg in stearic acid and caproic acid, as a function of concentration. Similar experiments with the polyglycolic acid immersed in distilled water produced no fracture after 24 h. These results confirm that these lipids have a very rapid effect on the polyglycolic acid, which could explain the loss of strength upon implantation.

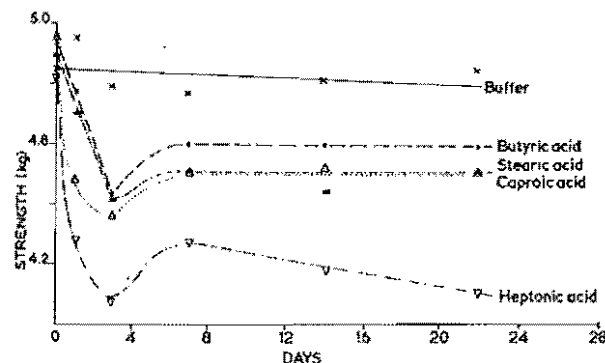


Figure 3 Loss of strength of polyglycolic acid after incubation in lipid solutions [48].

The re-implanted specimens have suffered slightly more degradation (55.8% strength remaining compared to 59.9%) which was statistically significant at $p < 0.005$, but which was hardly a conclusive difference. This was probably due to the fact that the time period is too short for there to be a significant change in the nature of the inflammatory tissue and tends to confirm the observation that the propagation phase of this hydrolysis is largely independent of the nature of the environment. The attention of the author has now been turned to other polymers of a more prolonged degradation profile which hopefully will give better results with the technique.

Enzymes, of course, are only one of several species of the physiological environment which could affect polymers, and it is necessary to consider the alternatives. One group of such alternatives are the lipids present in extra-cellular fluids. It is well-known that lipids are able to influence the structural integrity of some polymeric materials, most notably including silicone rubber [44-47]. Heart valves in which the ball was made of this material have malfunctioned because of structural changes in the rubber and a number of silicone rubber joint prostheses have fractured, apparently because of environmental-mechanical interactions. Although the exact reasons for this degradation are not entirely clear, it does seem certain that lipids and other highly polar substances are involved.

Some *in vitro* experiments were therefore performed in which polyglycolic acid sutures were exposed to varying concentrations of the series of lipids: butyric acid, caproic acid, heptonic acid and stearic acid [48]. Fig. 3 shows the very significant loss of strength of samples during the first few days. Interestingly, samples recover some of the strength and suffer little further degradation

over a period of time in these particular solutions (in a phosphate buffer at pH 7.0). This subsequent rise in strength may be associated with an absorption of lipids into the polymer or associated swelling. Fig. 4 shows the relationship between the time taken for a suture to break at a nominal load of 2.55 kg in stearic acid and caproic acid, as a function of concentration. Similar experiments with the polyglycolic acid immersed in distilled water produced no fracture after 24 h. These results confirm that these lipids have a very rapid effect on the polyglycolic acid, which could explain the loss of strength upon implantation.

4.4. *In vivo* degradation of "non-absorbable" sutures [41, 49]

Sutures are conveniently classified into those which are absorbable, i.e., dissolving in body tissues within a few months, and those which are non-absorbable. The difference is only a matter of degree, however, and many non-absorbable sutures

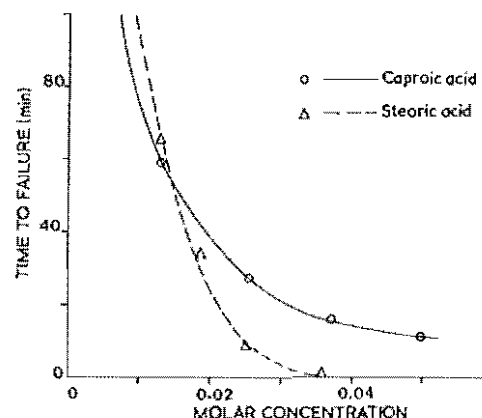


Figure 4 Time to failure of polyglycolic acid sutures immersed in caproic acid and stearic acids at an initial load of 2.55 kg, as a function of lipid concentration [48].

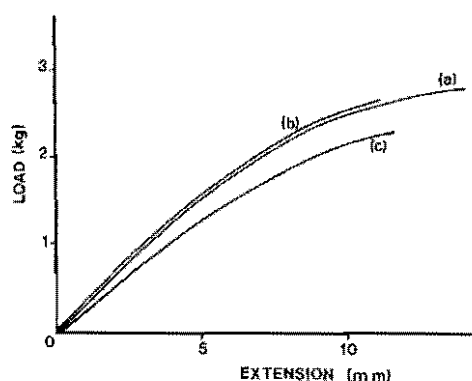


Figure 5 Load-extension curves for nylon: (a) control; (b) after 10 weeks implanted in one rat; and (c) after 10 weeks implanted 1 week in each of 10 rats [41].

do degrade slowly. These materials are, therefore, suitable for study by the re-implantation technique described above. Tests were performed with nylon (Ethilon[®], supplied by Ethicon Ltd.) and silk (Mersilk[®]). With the nylon, differences between specimens are obvious, as shown by the load-elongation curves given in Fig. 5. The specimens implanted in one rat for a total of ten weeks exhibited the same slope as the control, but fractured at a lower load and smaller elongation. Specimens continuously re-implanted every week for the same total of ten weeks, showed a smaller slope and a significantly lower breaking strength.

The curves for the silk sutures, given in Fig. 6, showed that the re-implanted samples gave much higher strengths than those maintained in single animals.

These results indicate that the continued presence of cells of the acute response are more important than cells of the chronic response in producing degradation of the nylon; while, with the silk, the situation is reversed.

Further tests have been carried out with the nylon in which the total period of implantation was 30 weeks with a re-implantation time of 3 weeks [49]. The results of the breaking strength showed the same trend as before, with a mean loss of strength of 11.3% in the re-implanted material and 5.7% in the single-animal experiments. The really interesting observation, however, is that the amount of degradation here is only marginally greater than that observed at 10 weeks, which amounted to just less than 10% in the re-implantation case and 2 to 3% in the single-

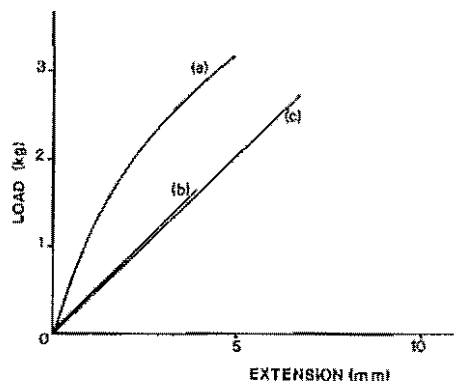


Figure 6 Load-extension curves for silk: (a) control; (b) after 10 weeks implanted in one rat; and (c) after 10 weeks implanted 1 week in each of 10 rats.

animal case. This would confirm the hypothesis that much of the degradation is taking place very soon after implantation and that little further degradation occurs thereafter. The significant degradation takes place during one week of implantation, but extending this period to 3 weeks produces little additional effect.

4.5. The effect of bacteria on absorbable sutures [50]

As a final commentary on the role of the physiological environment on polymer degradation, some experiments concerning the effects of bacteria on absorbable suture materials are worth reviewing. It is known that some bacteria are capable of degrading certain non-proteinaceous macromolecular structures, principally through the action of intracellular enzymes [51]. This would suggest that, if polyglycolic acid sutures were degraded by enzymes, then sutures in infected wounds might be more susceptible to degradation than those in clean wounds. Experiments were, therefore, performed to test this hypothesis, using cat-gut sutures for comparison.

Tests were carried out both *in vitro* and *in vivo*. In the former case, cat-gut and Dexon sutures were placed in broths at 37° C containing *Streptococcus mitis*, *Escherichia coli* or *Staphylococcus albus*. Table I shows the results of the breaking strength of sutures after 3 weeks, comparing the effects of incubation in both broths and bacterial cultures. It can be seen here that, while the cat-gut showed no difference, the polyglycolic acid sutures degraded more in the presence of the broth alone than when bacteria were present.

TABLE I Breaking loads of sutures:

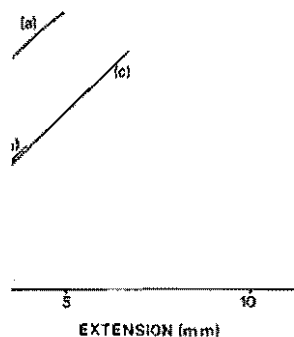
Suture	Bacteria
Cat-gut	<i>Streptococcus mitis</i>
	<i>Escherichia coli</i>
	<i>Escherichia coli</i>
	<i>Staphylococcus alb</i>
PGA	<i>Streptococcus mitis</i>
	<i>Escherichia coli</i>
	<i>Escherichia coli</i>
	<i>Staphylococcus alb</i>

These results were confirmed when the broths were artificial into account the lowering of

In the *in vivo* experiment implanted subcutaneously in sites being inoculated with *St*. Several experiments were performed the bacterial count and the pH were varied. The conclusion bacteria in sufficient number degradation of polyglycolic that polyglycolic acid and catgut different behaviour in this former degraded faster in the cat-gut degraded faster in infected the bacterial count was sufficient indicates a clear superiority absorbable suture in infected to the earlier observation by The reason for the observed glycolic acid are not clear at obviously the expected role of enzymes has not materialized.

5. Conclusions

This review of the literature experimental work has shown that polymers in the physiological show some differences to different environments. Some polymers expected to be quite stable under these conditions, while unstable polymers show a degradation profile. It is clear are able to influence the degradation polymers and there is some evidence to implicate lysosomal enzymatic degradation. Other factors, such as lipids or other organic species



ion curves for silk: (a) control; implanted in one rat; and (c) after week in each of 10 rats.

would confirm the hypothesis degradation is taking place very tation and that little further s thereafter. The significant place during one week of extending this period to 3 le additional effect.

f bacteria on absorbable

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TABLE I Breaking loads of sutures after incubation in broths and bacterial cultures [50]

Suture	Bacteria	Broth	Breaking strength (kg)		
			Control	In broth	In culture
Cat-gut	<i>Streptococcus mitis</i>	Tryptone soya	3.41	2.34	2.25
	<i>Escherichia coli</i>	Peptone-glucose	2.19	1.70	1.81
	<i>Escherichia coli</i>	Brain-heart	3.38	2.67	2.37
	<i>Staphylococcus albus</i>	Brain-heart	2.85	1.65	2.09
PGA	<i>Streptococcus mitis</i>	Tryptone soya	4.77	1.31	2.79
	<i>Escherichia coli</i>	Peptone-glucose	4.20	2.06	3.01
	<i>Escherichia coli</i>	Brain-heart	4.55	2.74	3.27
	<i>Staphylococcus albus</i>	Brain-heart	4.63	0.59	0.96

These results were confirmed in additional tests when the broths were artificially acidified to take into account the lowering of the pH by bacteria.

In the *in vivo* experiments, sutures were implanted subcutaneously in rats, half of the sites being inoculated with *Staphylococcus albus*. Several experiments were performed in which the bacterial count and the period of implantation were varied. The conclusion, again, was that bacteria in sufficient numbers inhibited the degradation of polyglycolic acid. It was clear that polyglycolic acid and cat-gut sutures displayed different behaviour in this respect. While the former degraded faster in the absence of bacteria, cat-gut degraded faster in infected tissue, provided the bacterial count was sufficiently high. This indicates a clear superiority for the synthetic absorbable suture in infected tissues, in contrast to the earlier observation by Sebeseri *et al.* [52]. The reason for the observed effects with polyglycolic acid are not clear at this stage, although obviously the expected role of bacterial-produced enzymes has not materialized.

5. Conclusions

This review of the literature and recent experimental work has shown that the degradation of polymers in the physiological environment may show some differences to degradation in other environments. Some polymers which would be expected to be quite stable do degrade slowly under these conditions, whilst hydrolytically unstable polymers show variations in their degradation profile. It is clear now that enzymes are able to influence the degradation of synthetic polymers and there is some circumstantial evidence to implicate lysosomal enzymes in *in vivo* degradation. Other factors, such as the presence of lipids or other organic species, have to be taken

into account, however, and more work needs to be carried out in order to clarify this phenomenon.

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A comparison of strength and

B. J. PLETKA*, S. M. WILSON
National Bureau of Standards

Predictions of lifetime under fracture mechanics techniques using fracture mechanics data, with confidence limits for the failure stressing-rate and fracture rate propagation and design applications.

1. Introduction

The utilization of ceramic materials in applications is often hampered by the brittle nature of these materials. A substantial variability in strength and dependence in strength known as scatter. Although it has been recognized that these effects of brittle behavior in small cracks or other flaws in ceramic materials, design techniques developed only recently to improve the reliability of these materials.

Design methods for ceramic materials are based on the premise that failure results from the growth of cracks that are present at the initial volume of the material. When the crack reaches a critical size, failure is instantaneous. The strength is determined primarily by the initial size of the crack that causes failure. The scatter that is observed in the strength is due to the scatter in the initial cracks present in the material. The time required for a component to fail under a load is the time necessary for the crack to grow from the initial size to the final size.

Techniques for predicting the lifetime of ceramic materials are based on the premise that failure results from the growth of cracks that are present at the initial volume of the material.

*Present address: Department of Materials Science, University of California, San Diego, 94395, USA.

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